Closed-loop Control of Functional Neuromuscular Stimulation

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This QPR is being sent to you before it has been reviewed by the staff of the Neural Prosthesis Program.

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1. SYNTHESIS OF UPPER EXTREMITY FUNCTION

The overall goals of this project are (1) to measure the biomechanical properties of the neuroprosthesis user's upper extremity and incorporate those measurements into a complete model with robust predictive capability, and (2) to use the predictions of the model to improve the grasp output of the hand neuroprosthesis for individual users.

1. a. BIOMECHANICAL MODELING: PARAMETERIZATION AND VALIDATION

Purpose

In this section of the contract, we will develop methods for obtaining biomechanical data from individual persons. Individualized data will form the basis for model-assisted implementation of upper extremity FNS. Using individualized biomechanical models, specific treatment procedures will be evaluated for individuals The person-specific parameters of interest are tendon moment arms and lines of action, passive moments, and maximum active joint moments. Passive moments will be decomposed into components arising from stiffness inherent to a joint and from passive stretching of muscle-tendon units that cross one or more joints.

Report of progress

1. a. i. MOMENT ARMS VIA MAGNETIC RESONANCE IMAGING

Abstract

In this quarter, we have again greatly improved image quality of finger images by improving the acquisition and the hand and finger holder. We have now analyzed the tendon moment arm on two subjects. In subject KJ, we find excellent agreement between the 3D geometric method and the tendon excursion method. In the case of subject RM, there is more difference between the two methods. Activity in the next quarter will focus on evaluating the accuracy and reproducibility of these measurements.

Progress Report

We are continuing to create methods for measuring tendon moment arm in the MCP joints of the fingers. As described in a previous report, we consider this joint to be simpler than the wrist; hence, we elect to study it first. We use high-resolution, 3D MRI to measure tendon moment arm, and our initial goal is to determine an accurate, practical method. As described in the proposal, we will examine at least 3 methods for analyzing tendon moment arm. They are: tendon excursion, 3D geometric, and 2D geometric. In two previous reports, we detailed the tendon excursion and the 3D geometric method. We are still working on the 2D method.

We are using a new MRI open magnet system which allows much easier patient access and hand manipulations. (A conventional closed magnet system requires that a subject lay down with her arm over her head.) A downside is that we have been required to develop new MRI sequences which optimize image quality on this 0.2 Tesla machine. Although we have been acquiring images for some time, in this quarter, we have had some great improvements in image quality. We are now using a larger coil with the hand running through the center of the coil. Optimized acquisition consists of a TR = 40 ms and a TE = 12 ms. We also increased the stability of our hand and finger holder. This reduced motion blur of the subject as well as subject fatigue.

We have now analyzed the moment arm of the profundus tendon of the right hand 3rd MCP joint in two subjects. In the case of the first subject, images were acquired at 4 positions including the neutral position. Segmentations and joint center analyses are remarkably consistent. Measured moment arms are 12.45, 14.9, and 13.25 mm for joint angle pairs of (0.0-19.31 deg), (19.31-30.17 deg), and (30.17-50.93 deg). In the case of the second subject, images were acquired at 4 positions including the neutral position. Segmentations and joint center analyses are consistent for the two measurement methods. Measured moment arms were 5.95, 15.60, and 13.03 mm for joint angle pairs of (21.66-35.89 deg), (35.89-49.20 deg), and (49.20-61.21 deg). The percent difference between the averaged geometric method and the smoothed tendon excursion method is 43%, 32%, and 34%. The consistency within the first subject was remarkable. The consistency within the second subject was not as good as the first. Activity in the next quarter will be focused on understanding the accuracy of these data.

Plans for next quarter

We will repeat the measurements on subject 2. We will acquire a new image data set and analyze it. This should give us an idea of the reproducibility. We are also planning to do some computer and physical phantom validations. In both cases, we will know the center of rotation and angle changes involved. We will test our ability to make this measurement. Noise will be added to simulations to test the propagation of segmentation error.

1.a.ii. PASSIVE AND ACTIVE MOMENTS

Abstract

During the past quarter, a sensitivity analysis of the derived joint parameters described in the previous report was done. Additional passive moment measurements were made on the index finger metacarpophalangeal (MP) joint of both able-bodied and tetraplegic individuals. Data collected with the wrist in three positions were examined for trends. The three wrist positions were 0°, 60° of extension, and 60° of flexion. EMGs of the extensor communis and flexor digitorum superficialis were monitored during the rotation of the MP joint to verify passivity in one normal and one paralyzed subject.

Purpose

The purpose of this project is to characterize the passive properties of normal and paralyzed hands. This information will be used to determine methods of improving hand grasp and hand posture in FES systems.

Report of Progress

Review of Parameter Definitions

The previous report described how six parameters were derived from the moment-angle curves (MACs) that were measured from subjects. Four of these parameters are slopes of line segments that connect certain points on a curve fitted to each MAC. Since these are slopes of a moment-angle curve, they give an indication of the stiffness of the joint as it moves through its range. These parameters cannot strictly be called the stiffness of the joint because they are not derivatives of the moment-angle curves. Nevertheless, they do give us an indication of the joint stiffness that can be compared across subjects. Two of these parameters indicate the stiffness of the joint as it extends and two correspondingly indicate the stiffness as the joint flexes. The parameters are labeled as Ext(0-20), Flx(0-20), Ext(20-30), Flx(20-30). The remaining two parameters derived from the MACs are the passive range of motion of the joint (PROM) and the rest position of the joint, Ør. PROM is defined in this study as the joint rotation range between the bounds of -20 and 20 N·cm. The rest position is a parameter that directly results from the curve fit. It is the joint angle at which the curve fit crosses the abscissa (where the passive moment is zero), and is therefore a measure of the angle at which the MP joint rests for the given wrist position.

Sensitivity Analysis of Derived Parameters

The repeatability of the apparatus in measuring the passive MACs was evaluated. The first repeatability experiment measured the consistency of the MAC parameters when the hand is not removed from the apparatus between measurements. In a second repeatability experiment, the hand is removed from the apparatus and then remounted between measurements. Coefficients of variation (standard deviation divided by mean) were calculated for each derived parameter. The coefficients of variation for the four stiffness parameters ranged from 0.3% to 1.4% in the test where the hand was not removed from the apparatus and ranged from 3.3% to 4.6% in the test when the hand was removed from the apparatus and remounted between measurements. PROM had a coefficient of variation of 0.4% in the first experiment and 3.0% in the second, while the rest position had coefficients of variation of 1.1% and 9.3% in the respective experiments.

The repeatability tests only take into account how realigning the apparatus to the joint and resplinting the finger affect the measured parameters. They do not tell us anything about how much these parameters change for a single individual over time (days, weeks, or months). The previous report addressed the repeatability of the MAC over five weeks. Curves were fit to that data and from the resulting curve fit parameters the derived stiffness parameters, PROM, and rest position were determined for each MAC. Again, coefficients of variation were computed for each derived parameter. The coefficients of variation for the four stiffness parameters ranged from 3.2% to 10.8%. PROM had a coefficient of variation 3.2%, while the rest position had a coefficients of variation of 17.6%.

MACs Measured With the Wrist Fixed at Three Position

During this quarter, additional index finger MP joints were tested with the wrist fixed in different positions. The total number of subjects now tested under these conditions is 2 patients with tetraplegia, and 6 normal controls. The passive moments measured when the wrist was fixed at 0°, 60° of extension, and 60° of flexion were compared. For each MAC the four stiffness parameters (Ext(0-20), Ext(20-30), Flx(0-20), Flx(20-30)) as well as rest position and PROM were computed for each wrist position. The average values for these parameters were used to compare the general passive properties between paralyzed and normal subjects.

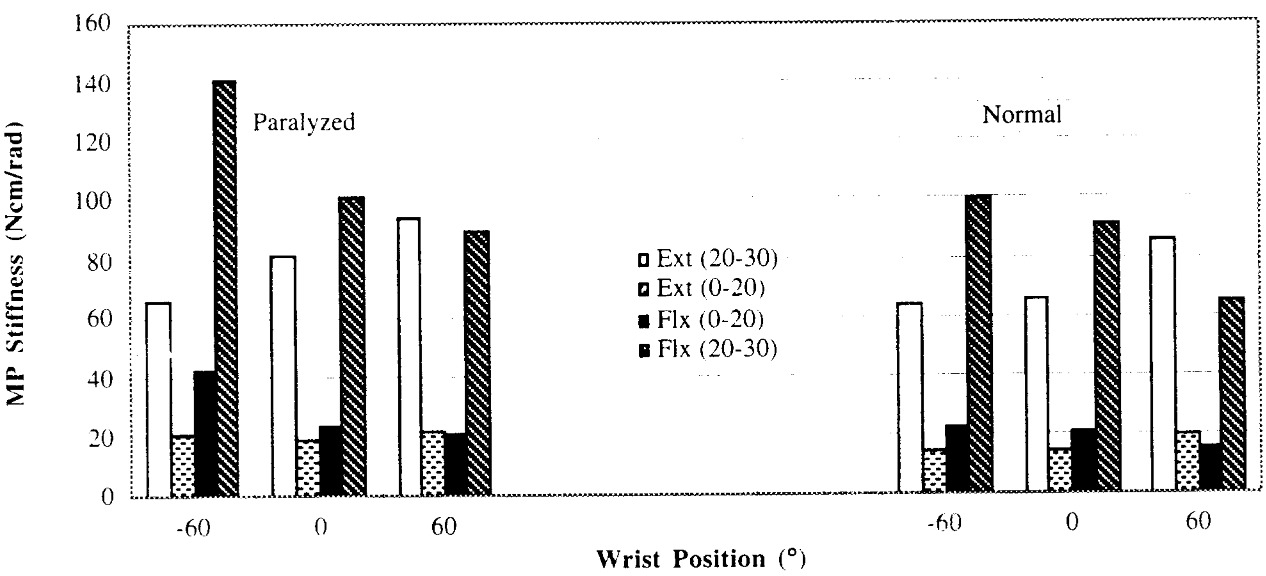


Figure 1.a.ii.1. Average MP stiffness parameters at each wrist position for paralyzed and normal subjects. For paralyzed subjects, n=2. For normal subjects, n=6 when wrist position is -60° or 0°, but n=5 when wrist is 60° flexed.

Figure 1.a.ii.1 shows the average stiffness parameters at each of the three wrist positions. At each of the three wrist positions the average magnitude of each stiffness parameter is greater for the paralyzed subjects than for the normal controls. This figure also shows that when the wrist is straight or extended, the stiffness values were greater in the flexion region of the MAC than the corresponding stiffness values in the extension region. However, with the wrist flexed, the flexion stiffness values were less than or nearly equal to the corresponding extension stiffnesses for both the paralyzed and normal populations. Therefore, in general, as the wrist rotates from extension to flexion the extension side of the MAC gets steeper (stiffer) while the flexion side of the MAC gets flatter (more compliant). Furthermore, as far as the stiffness parameters are concerned, this figure shows that the average normal's MP joint behaves more symmetrically about the 0° wrist position than the average paralyzed person's MP joint. That is, the magnitudes of the extension and flexion stiffnesses for the extended wrist nearly mirror the magnitudes of

the flexion and extension stiffnesses for the flexed wrist for the average normal, but this is less apparent for the average paralyzed person's joint. For the average paralyzed person, the flexion stiffnesses are still nearly equal to the extension stiffnesses even with the wrist flexed. This suggests that the tetraplegic MP joint is more "biased" in flexion than the normal MP joint. The flexors are perhaps tighter in the paralyzed hand than in the normal hand. More subjects must be studied to verify these preliminary findings.

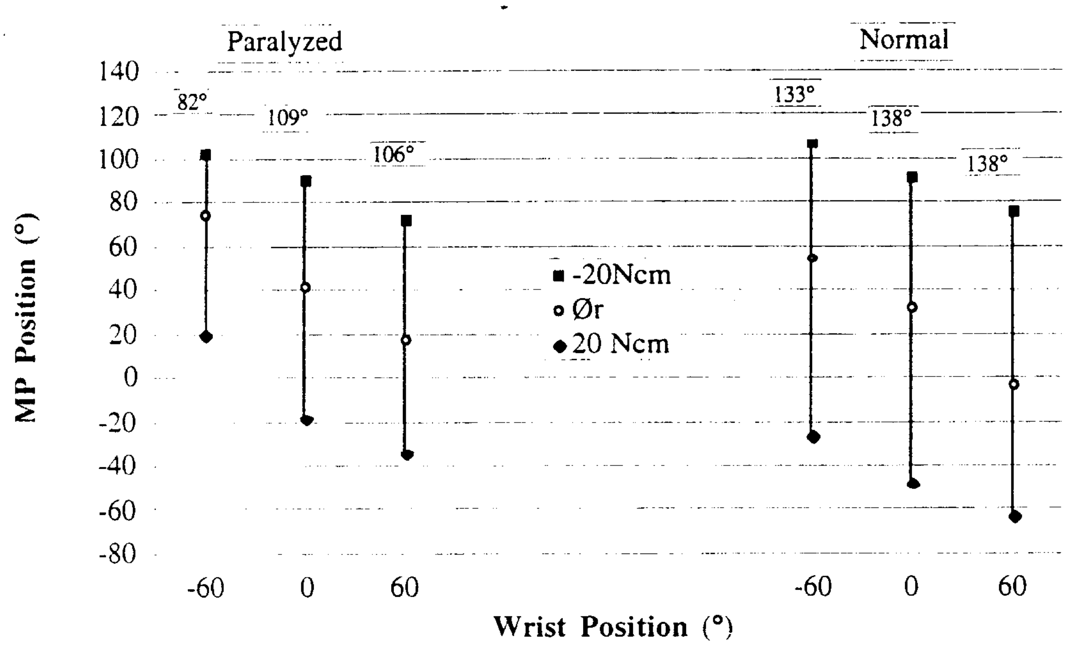


Figure 1.a.ii.2. Average PROM and rest position at each wrist position for paralyzed and normal subjects. For paralyzed subjects. n=2. For normal subjects n=6 when the wrist position is -60° or 0°. but n=5 when wrist is 60° flexed. Extension is positive and flexion is negative.

Figure 1.a.ii.2 shows the average PROM and rest position (Ør) at each of the three wrist positions. As the wrist is flexed, the rest position of the MP joint becomes more extended and the entire range of motion is shifted extensionward. However, in each wrist position the average tetraplegic's joint rested in a more flexed position than the average normal's joint. The PROM did not change much as a function of wrist position for the normal joint, but for the paralyzed hand the PROM significantly decreased as the wrist extended from 0° to 60°. At each of the three wrist positions PROM is less for the average paralyzed subject than the PROM for the average normal subject. The average paralyzed subject could not extend the MP as far as the average normal in each of the wrist positions.

EMG Studies

Surface electrodes were used to study the EMGs of the extensor communis and flexor digitorum superficialis while the index finger is rotated through its passive range using our apparatus. One patient with tetraplegia and one normal control were studied. The EMGs from both muscles of both subjects remained quiescent during the finger rotation.

Plans for Next Quarter

Next quarter the index MP joints of more tetraplegic and normal subjects will be tested with the wrist fixed in the three positions. Data analysis will continue to identify if there are indeed quantitative differences between the passive moments of joints of paralyzed and normal hands and to explain how those differences manifest themselves.

Reference

Esteki, A. and Mansour, J. M. (1996) An experimentally based nonlinear viscoelastic model of joint passive moment. *J. Biomechanics* 29, 443-450.

1. b. BIOMECHANICAL MODELING: ANALYSIS AND IMPROVEMENT OF GRASP OUTPUT

Abstract

This project does not start until year three of the project, as described in the proposal. However, two important tools are being developed in order to make the necessary biomechanical measurements with individual patients. First, the use of magnetic resonance images to determine joint moment arms is described in Section 1.a.i. Secondly, the measurement of passive moments across all joints of the hand is described in Section 1.a.ii.. When these tools are complete, we will begin making measurements on both normal and paralyzed patients.

Objective

The purpose of this project is to use the biomechanical model and the parameters measured for individual neuroprosthesis users to analyze and refine their neuroprosthetic grasp patterns.

2. CONTROL OF UPPER EXTREMITY FUNCTION

Our goal in the five projects in this section is to either assess the utility of or test the feasibility of enhancements to the control strategies and algorithms used presently in the CWRU hand neuroprosthesis. Specifically, we will: (1) determine whether a portable system providing sensory feedback and closed-loop control, albeit with awkward sensors, is viable and beneficial outside of the laboratory, (2) determine whether sensory feedback of grasp force or finger span benefits performance in the presence of natural visual cues, (of particular interest will be the ability of subjects to control their grasp output in the presence of trial-to-trial variations normally associated with grasping objects, and in the presence of longer-term variations such as fatigue), (3) demonstrate the viability and utility of improved command-control algorithms designed to take advantage of forthcoming availability of afferent, cortical or electromyographic signals, (4) demonstrate the feasibility of bimanual neuroprostheses, and (5) integrate the control of wrist position with hand grasp.

2. a. HOME EVALUATION OF CLOSED-LOOP CONTROL AND SENSORY FEEDBACK

Abstract

The purpose of this project is to deploy an existing portable hand grasp neuroprosthesis capable of providing closed-loop control and sensory feedback outside of the laboratory. Effort this quarter was devoted to preliminary laboratory testing of the portable feedback system and the evaluation task with four hand neuroprosthesis users. The compliant object was modified, reducing its weight by 87%, and was glued to the thumb-mounted sensor to obviate the need for subjects to manipulate (rather than simply squeeze) the object. Some of the subjects completed the task with or without vision, but others had difficulty. The preliminary results suggest that sensory feedback of grasp force can assist users in acquiring and maintaining a fixed grasp force.

Purpose

The purpose of this project is to deploy an existing portable hand grasp neuroprosthesis capable of providing closed-loop control and sensory feedback outside of the laboratory. The device is an augmented version of the CWRU hand neuroprosthesis, and was developed and fabricated in the previous contract period. The device utilizes joint angle and force sensors mounted on the thumb to provide sensory information, and requires daily support from a field engineer to don and tune. The portable closed-loop system (PCLS) is not intended as a long term clinical device. Our goal, rather, is to evaluate whether the additional functions provided by this system benefit hand grasp outside of the laboratory, albeit with poor cosmesis and high demands for field support.

Report of Progress

Two important objectives were accomplished this quarter. First, the PCLS was tested in day-long trials by two subjects: one able-bodied, and one with tetraplegia. The latter was performed as a physical mock-up in that all components of the grasp-force sensory feedback configuration were donned, though the system was not active. Prior to the day-long trial, we did confirm that the PCLS could control this user's neuroprosthesis in the laboratory. The initial passive trial was essential for assuring that the sensor (mounted on a thumb-pick and worn on the thumb) and its cable did not impede the user's grasp and did not produce irritation of the digit. The subject reported that the sensor did not interfere with his grasp, even during hand intensive activities such as eating. He did report, though, that the sensor did not always contact the object being grasped. The trial with the able-bodied subject again used the force sensory feedback configuration, with feedback active, but without muscle stimulation. The subject wore the thumb sensor, feedback electrode (placed at the base of the neck over the superior aspect of the trapezius muscle) and carried the control unit for approximately 6 hours before the battery discharged. The feedback was active for the entire duration of the trial. The sensor caused no irritation of the thumb, nor did it significantly impede hand use during normal activities including office tasks and eating. Some effort was required to position the sensor optimally by fine movements of the thumb, confirming early expectations. Neuroprosthesis users will not have such dexterity, so the sensor will not work well all the time. The feedback stimulation was stable subjectively throughout the day, and was comfortable. Stimuli generated by tasks that were repeated frequently, such as grasping a pen, produced reliable sensations. Moreover, the continuous feedback yielded perceptions of expectation; that is, the subject reported expecting (though not relying on) feedback sensations of particular magnitude during familiar tasks.

The second accomplishment was completion of the software changes mentioned in the previous report. Changes included: (1) use of either palmar or lateral sensors with either grasp, (2) improved navigation between program windows, (3) more flexible programming of maximum and minimum forces and spans, and (4) corrected calibration procedure.

Last, a hardware failure prevented additional field testing. One of the voltage sources failed leading to erroneous battery discharge indication and system shut-down. The power systems are being evaluated.

Plans for Next Quarter

We will complete at least two day-long tests using able-bodied subjects, and will continue to search for a neuroprosthesis user to initiate field trials.

2. b. INNOVATIVE METHODS OF CONTROL AND SENSORY FEEDBACK

2. b. i. ASSESSMENT OF SENSORY FEEDBACK IN THE PRESENCE OF VISION

Abstract

The purpose of this project is to develop a method for including realistic visual information while presenting other feedback information simultaneously, and to assess the impact of feedback on grasp performance in the presence of such visual information. Progress on this project was delayed. The student assigned to the project was forced to withdraw due to illness. A new student has been recruited and is in the process of reviewing all previous work on the project.

Purpose

The purpose of this project is to develop a method for including realistic visual information while presenting other feedback information simultaneously, and to assess the impact of feedback on grasp performance. Vision may supply enough sensory information to obviate the need for supplemental proprioceptive information via electrocutaneous stimulation. Therefore, it is essential to quantify the relative contributions of both sources of information.

Report of Progress
(see Abstract)

Plans for Next Quarter
The complete simulation system will be completed and tested on the new computer, as proposed originally for the previous quarter.

C. 2. b. ii. INNOVATIVE METHODS OF COMMAND CONTROL

Abstract
During this quarter we conducted experiments with two upper extremity neural prosthesis users to test the performance of a strain gage mounted on the thumb nail as contact sensor. The output of a strain gage was recorded while the subjects performed functional reaching tasks with an instrumented object. The results indicated that the gage output was correlated with contact and grasp force. However, the signals were corrupted by sensitivity of the strain gage to elevation of the hand. During this quarter we also developed software routines that enable communication with the neural prosthesis control unit via LABView software of the laboratory computer. This software will enable us to acquire, process, and re-issue to the output processor the user's command signals as required for evaluation of new command control algorithms.

Purpose
The purpose of this project is to improve the function of the upper extremity hand grasp neuroprosthesis by improving user command control. We are specifically interested in designing algorithms that can take advantage of promising developments in (and forthcoming availability of) alternative command signal sources such as EMG, and afferent and cortical recordings. The specific objectives are to identify and evaluate alternative sources of logical command control signals, to develop new hand grasp command control algorithms, to evaluate the performance of new command control sources and algorithms with a computer-based video simulator, and to evaluate neuroprosthesis user performance with the most promising hand grasp controllers and command control sources.

The first objective is to identify and evaluate alternative sources of logical command control signals. We will investigate object contact and object slip detection using sensors mounted on the dorsal surface of the thumb. The first sensor investigated was a strain gage glued to the thumbnail and used to detect object contact.

Report of Progress
During this quarter we conducted 2 experiments to evaluate the performance of thumbnail-mounted strain gages as contact detectors.

METHODS
All subjects read and signed an informed consent, and all procedures were reviewed by the Institutional Review Board of MetroHealth Medical Center.

A metal foil strain gage (SG-3/350-LY13, Omega Engineering, Stamford, CT)) was glued to the thumbnail of the dominant hand approximately perpendicular to the long axis of the thumb, using cyanoacrylate cement. A custom-built instrumentation amplifier, based around the Analog Devices 1B31AN chip, was used to provide excitation voltage (5V), amplification, and low-pass filtering (25 Hz) of the strain gage signals. One of the two subjects (CWJ) was also instrumented with a large carbon rubber surface electrode (6282, 3M Health Care, St. Paul, MN) that formed 1 pole of a continuity detector. The other pole of the continuity detector was formed by conductive foil placed on one side of the test object. A custom-built detector generated a signal when the hand-object circuit was completed and was used to provide an independent measure of when the thumb contacted the object to be grasped. The object use in these trials was the instrumented "book" which provides a voltage output proportional to the grasp force. [Memberg and Crago, 1997]. The test object was placed on a table switch that generated a TTL signal when the object was lifted off the table.

Each trial consisted of the subject reaching out, grasping, lifting the object, transporting it to another location, and releasing the object. The beginning of a trial was signaled by an audible "GO" signal from the experimenter. This first subject (JHJ) performed 68 trials and the second subject (CWJ) performed 102 trials with the object at different locations in the workspace and in different orientations.

RESULTS

Our previous results indicated that the output of the strain gage was affected by the position of the thumb, as well as contact of the thumb with the object to be grasped (see QPR #2). During the present experiments very flexible coiled lead wires were used between the gage and a cable on the dorsal surface of the hand. This arrangement reduced greatly the gage sensitivity to thumb position.

Elevation of the hand also had a significant effect on the output of the gage. In previous experiments a "horizontal" task of lifting, holding and replacing an object in a single location was used (see QPR #2). The total elevation change of the object was only ~10 cm. In contrast, the task used in the present evaluations required functional reaching and included elevation changes of ~50 cm. These changes in hand elevation generated large changes in the output of the gage which were superimposed on the thumb contact and grasp force signals.

The results with the first subject proved somewhat difficult to interpret because no positive contact signal detector was used. However, there was a strong correlation between an increase in the grasp force registered on the instrumented book and the output signal of the nail mounted strain gage. The gage output and grasp force during two representative trials are shown in fig. C.2.b.ii.1. In these trails the subject grasped the object at a "low" location and moved it to a "high" location. The initial decrease in gage output (* in fig. C.2.b.ii.1) corresponded to the subject reaching toward the instrumented book position on a table at the height of his abdomen. This was followed by an increase in gage output corresponding to contact (arrow) and subsequent force application to the object. Note that the increase in force was also registered by the instrumented book. The output of the instrumented book preceding this increase in force resulted from the subject placing his fingers on the object with his hand open ("f" in fig. C.2.b.ii.1). The strategy used by this subject to acquire the object was to contact the object first with his fingers, then close his hand grasp resulting in subsequent thumb contact and grasp of the object. During grasp the output of the strain gage tracks closely the grasp force as measured with the instrumented book. However, there is a large increase in strain gage output (** in fig. C.2.b.ii.1) which is not registered by the instrumented book. This apparent increase in grasp force resulted from the subject reaching up (i.e., elevating his hand) to place the instrumented book on a platform ~ 50 cm above the initial position. Thus, elevation related signals were superimposed on the strain gage output.

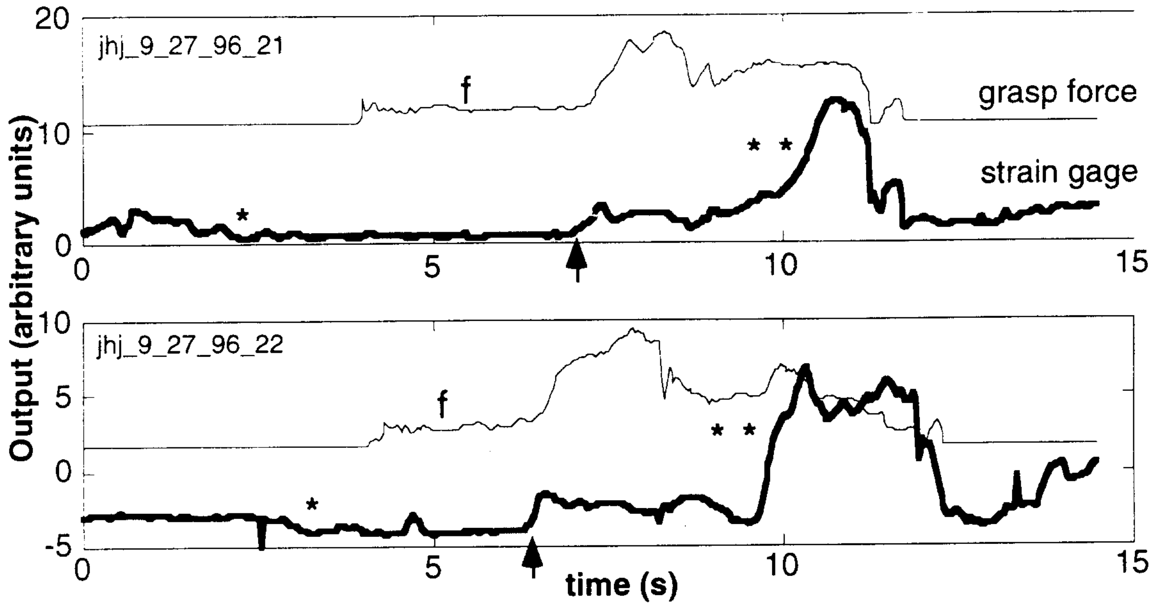


Figure C.2.b.ii.1: Records of the grasp force and nail mounted strain gage output during two trials in which the subject grasped the instrumented book at a low location, moved it to a high location, and released it. The traces have been offset for clarity.

Records of positive contact and lift-off of the instrumented book were made during the experiment with the second subject. Traces of these signals, the gage output, and the grasp force registered by the instrumented book during three representative trials are shown in fig. C.2.b.ii.2. In these trials the subject grasped the object at a "high" location and moved it to a "low" location. The initial increase in gage output (* in fig. C.2.b.ii.2) is a result of the elevation of the hand as the subject reached up to grasp the object. This is followed by an increase in output (arrow in fig. C.2.b.ii.2) as the thumb contacts the object and force was applied. The amplitudes of the contact and force signals are smaller than in our previous experiments because the gain of the strain gage amplifier had to be reduced to prevent saturation by the large elevation induced signals. The increased output of the strain gage in many cases preceded the output of the force sensor indicating thumb contact but very small force application to the instrumented book. The output of the strain gage tracks very well the grasp force registered by the book for the first part of the trial (t in fig. C.2.b.ii.2). The initial force tracking, however, is followed by a large shift in the strain gage output (m in fig. C.2.b.ii.2) which is not reflected in the output of the grasp force sensor. This shift resulted from a change in the elevation of the hand as the object was moved down from the elevated platform to the table top. After the transient, the strain gage output is again correlated with the output of the grasp force sensor. Thus, as seen in the first subject, the strain gage output was correlated with object contact and grasp force, but elevation related signals were superimposed on this output.

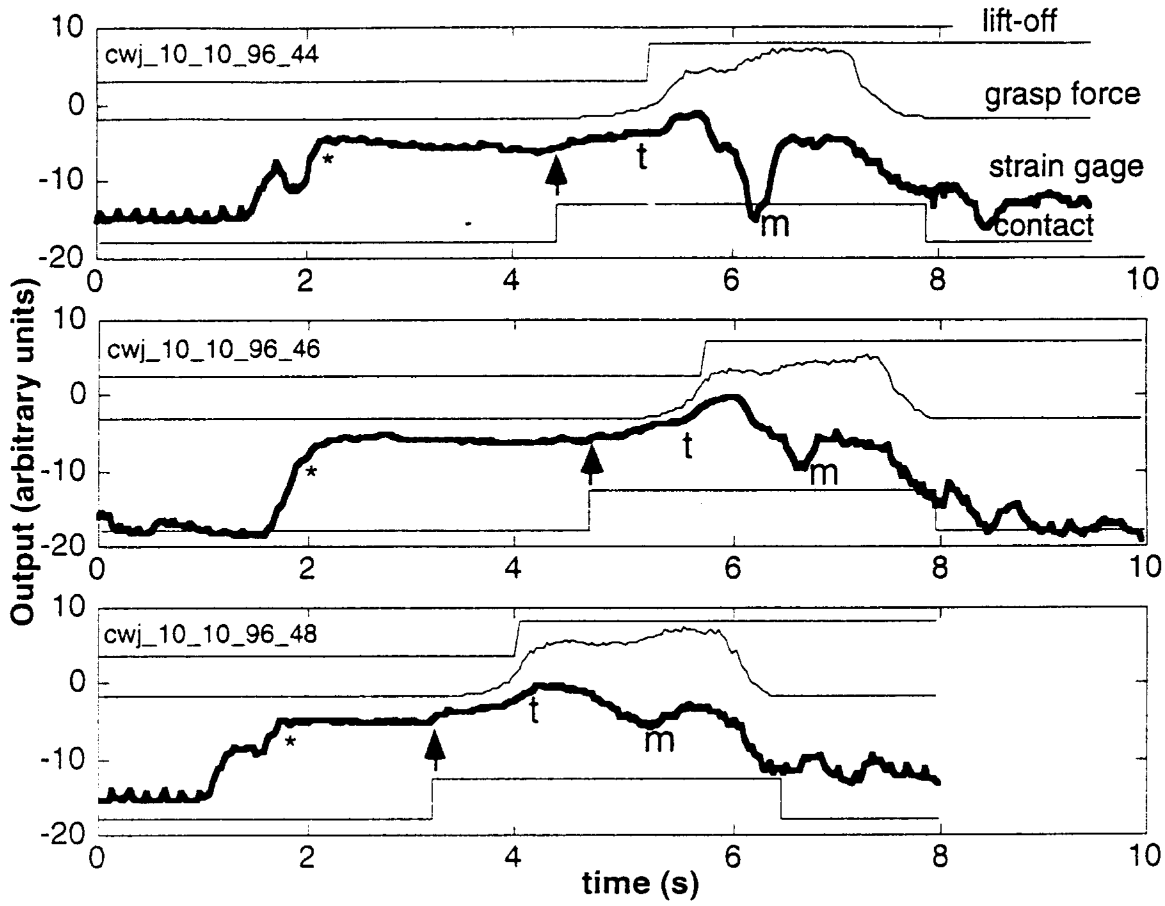


Figure C.2.b.ii.2: Records of object lift, grasp force, nail mounted strain gage output, and object contact during three trials in which the subject grasped the instrumented book at a high location, moved it to a low location, and released it. The traces have been offset for clarity.

SOFTWARE

During this quarter we also developed a software interface in LABView to allow monitoring, processing, and passing to the output processor of patient command control signals. This new software is required to implement and evaluate new command control algorithms, as well as to monitor command control signals during development of the computer-based video simulator.

The control algorithms in the upper extremity external control unit (ECU) were modified to include an additional flag. When the flag is set, proportional command data (from either the wrist controller or shoulder controller) and the machine state are sent over a serial communication line, rather than to the output processor. Additionally, the output processor is set to expect the command signals from a serial communication line, rather than from an internal source. The loop between command signal and stimulus output is de-coupled, and additional processing can be inserted into the loop.

A software module was written in LABView to enable the lab computer to monitor and record the incoming command signals, as well as to send out either the unprocessed or processed command signals to the output processor. In the absence of any command signal processing, the read and write of the command signal with a Mac IIfx introduces an additional delay of ~ 37 ms between the command and output processors in the ECU. This delay is not expected to affect patient command control because of the low bandwidth of the command signal [Hines et al., 1992], and can be reduced further using a faster computer.

Plans for Next Quarter

During the next quarter we will complete analysis of the data from the neural prosthesis users' trials. Specifically we will investigate the use of "Receiver Operator Characteristics" and different filtering options to optimize contact detection in the presence of contamination by the elevation signals. We will also conduct additional experiments to evaluate the use of multiple strain gages (rosette) on the thumbnail and differential measurements to isolate contact and grasp force information from elevation signals.

References

Hines, A.E., Owens, N.E., Crago, P.E. (1992) Assessment of input-output properties and control of neuroprosthetic hand grasp. IEEE Trans. Biomed. Eng. 39:610-623.

Memberg, W.D., Crago, P.E., "Instrumented Objects for Quantitative Evaluation of Hand Grasp", Journal of Rehabilitation Research and Development, vol. 34(1), 1997 (in press).

2. b. iii. INCREASING WORKSPACE AND REPERTOIRE WITH BIMANUAL HAND GRASP

Abstract

Studies were performed during this quarter on the recruitment properties of the extrinsic and intrinsic muscles of the hand. These studies were necessary to allow for the proper incorporation of the intrinsic muscles into the neuroprosthesis. Six subjects participated in this study, two of which had received the ten channel implant which provided electrical activation of the intrinsic muscles. Preliminary analysis of the data was completed, which demonstrated the strengthening effect of electrical stimulation over time as well as provided a quantitative measure of electrically activated muscle strength as it is related to the stimulus applied.

Purpose

The objective of this study is to extend the functional capabilities of the person who has sustained spinal cord injury and has tetraplegia at the C5 and C6 level by providing the ability to grasp and release with both hands. As an important functional complement, we will also provide improved finger extension in one or both hands by implantation and stimulation of the intrinsic finger muscles. Bimanual grasp is expected tc provide these individuals with the ability to perform over a greater working volume, to perform more tasks more efficiently than they can with a single neuroprothesis, and to perform tasks they cannot do at all unimanually.

Report of progress

In this quarter, studies were performed to provide quantitative information on the functioning of the individual hand muscles in neuroprosthesis users. Measurements were made on existing neuroprosthesis users as well as new neuroprosthesis recipients who had electrodes placed in the intrinsic (interosseous) muscles of the hand. This information can then be used to provide the user with the most functional hand grasp possible as well as provide information on muscle changes over time. This information is also valuable in order to allow for the incorporation of the intrinsic muscles into the hand grasp.

<u>Subjects</u>. Measurements were made on six neuroprosthesis users who ranged in age from 20 to 50 years. The subjects were varied in the number of years that they had been using their neuroprostheses. The most experienced user had been using his system for ten years while the newest user had only been using his system a few months.

<u>Finger Moment Transducer.</u> A device was developed and built to measure the moments generated about the metacarpophalangeal and interphalangeal joints of the hand (Kilgore, et al.). The design utilizes a parallelogram construction that allows the device to fit on different sizes of hands. Aluminum uprights, fitted with strain gages, were attached to each joint segment providing the measurement of isometric moments generated about the joints of a single digit. Ball bearing joints between the upright beams and the parallel crossbars prevent hysteresis as well as provide a linear relationship between the value measured by the strain gages and the applied moment. Each joint of the finger can be positioned and locked at any angle throughout the range of motion of the joint. The repeatability of the moment measurements was less than

0.57 and the error was less than +/- 1.5% in the +/- 100 N-cm range, the maximum range allowable for this device.

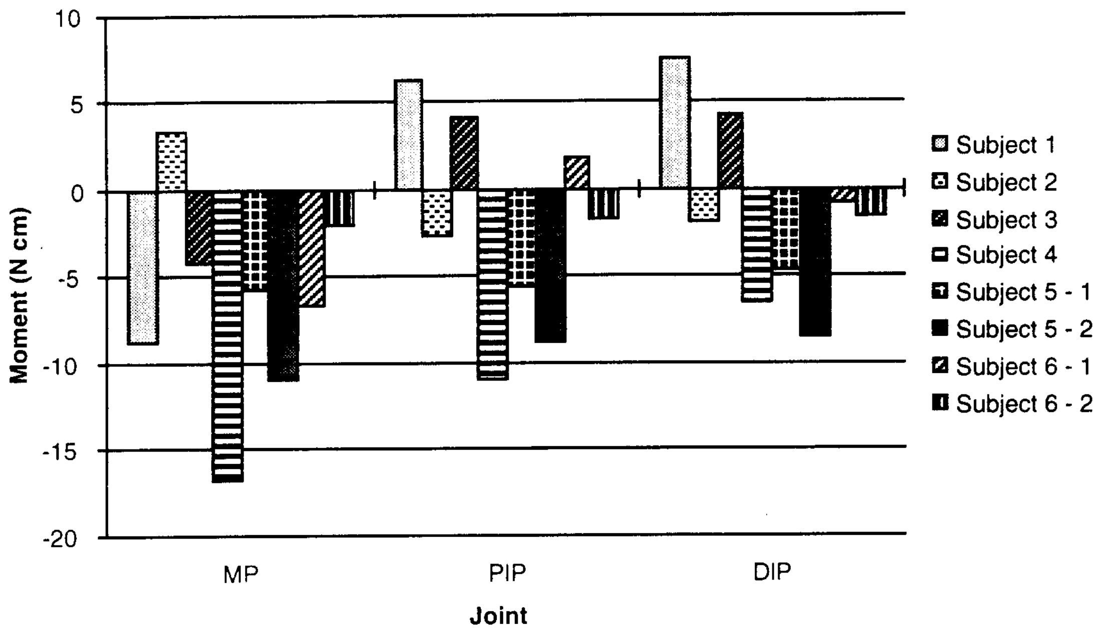
Additional Hardware. Stimulation of the individual muscles was accomplished using the implanted neuroprosthesis. Electrical activation of the muscles was controlled using the clinical interface software that is used to program the external control unit. Data collection was accomplished through a program written in LabVIEW running on a Macintosh IIfx computer. The analysis of the data was accomplished through the use of Microsoft Excel on the Macintosh.

Experimental Protocol. The subject was seated comfortably with the hand implemented with the neuroprosthesis resting on a raised platform. Each of the muscles to be studied were then profiled to determine the threshold level and the maximum stimulus level possible. The hand was then placed into the finger moment transducer and locked into an anatomically neutral position, or as close as possible depending upon the amount of flexibility that existed in the fingers. Using the profile information, the available range was broken down into a minimum of six levels of stimulation, evenly spaced throughout the available range. Each muscle was then stimulated at each level, and the moments across all the joints of the fingers were recorded. At least one minute of rest was given between each level of stimulation. The muscles studied were the extensor digitorum communis (EDC), the flexor digitorum profundus (FDP), the flexor digitorum superficialis (FDS), and the interossei of the fingers. Not all of these muscles were available in each user. This depended upon the extent of denervation due to the spinal cord injury, as well as the type of neuroprosthesis used (either the 8 channel or 10 channel implant).

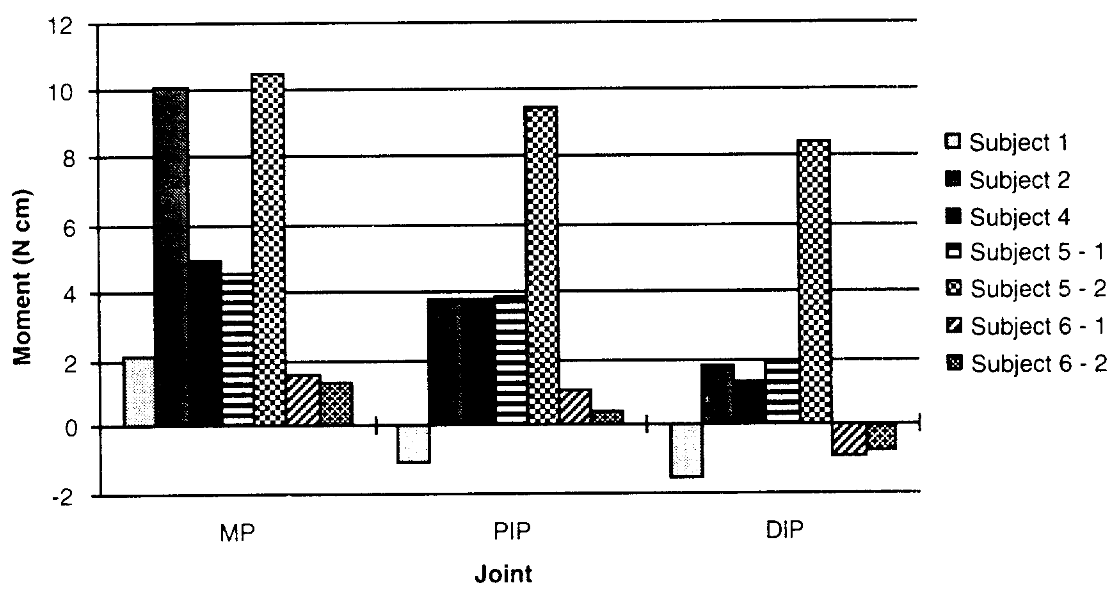
Moments generated by the finger flexors and extensors. The maximum moment measurements for the EDC and the FDS on the long finger are shown in Figure C.2.b.iii.1. Negative moments indicate flexion while positive moments indicate extension. Subject 1 does not have active finger extension since the tendons of the EDC are scarred down and no longer move freely. Instead, finger extension is achieved through a Zancolli lasso procedure which allows the FDS muscle to provide for interphalangeal joint extension. Extension of the metacarpophalangeal joint is provided through tenodesis to allow wrist flexion to generate extension at that joint.

Subjects 5 and 6 were the first subjects in which it was possible to record the finger joint moments over a period of time, starting from the time the neuroprosthesis was first used. The initial measurements are from one week after the casts on the arms are removed, while the second measurement is at least one month later. Subject 5 shows definite improvement in the muscles' moment generating capabilities, between 25 and 200%. Subject 6 only shows minor improvement in muscle strength, but definite improvement in muscle action. Further measurements on these subject are planned to observe stability of the muscles over time.

Moments generated by the interosseous muscle of the hand. The moments recorded for the second dorsal interosseous are shown in Figure C.2.b.iii.2. The measurements presented for Subject 1 were made during an initial study on the actions of the intrinsic muscles (Lauer, 1996). The moments measured in the neuroprosthesis users are shown in comparison to the moments recorded from able bodied subjects using electrical stimulation. Subject 1 was capable of generating moments that were only 9 to 13% of those generated by able bodied subjects. This is because the muscles in Subject 1 were deconditioned as a result of his paralysis. In comparison, Subjects 5 and 6 were capable of generating moments that were 50 to 90% of those generated by able bodied subjects.



Flexion - FDS Muscle



Extension - EDC Muscle

Figure C.2.b.iii.1 - Finger Flexor and Extensor Moments

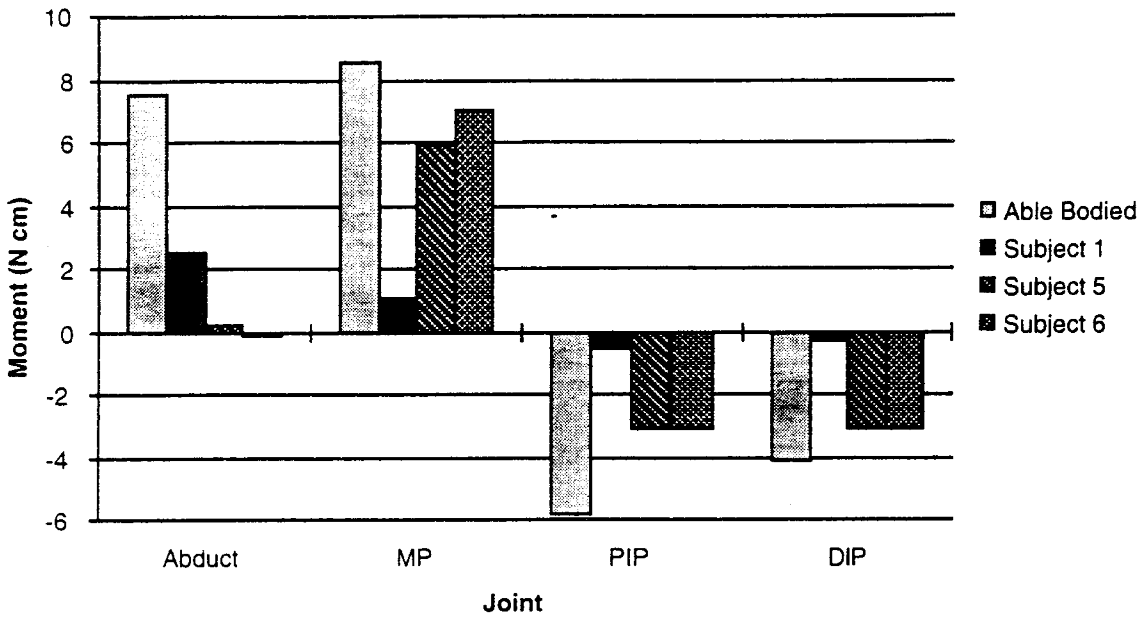


Figure C.2.b.iii.2 - Second Dorsal Interosseous Muscle Moments

The studies presented here provide direct measurements of the moment generating capabilities of electrically stimulated muscles in the neuroprosthesis users. These direct measurements have allowed us to monitor the effect of electrical stimulation on muscle strength over time. This will allow for modification and adaptation of the grasp templates as the muscles strengthen and change with use. In addition to following the effects on muscle strength over time, this data also allows for a comparison of the electrically conditioned paralyzed muscle to the healthy muscle. This will provide us with some measure of how much function a person using a neuroprosthesis can regain.

The measurement of joint moments provides the basis for automated synthesis of hand movement. For example, hand grasp can be optimized by the computer to set up the movement template so that the need for judgment and experience is reduced. In addition, the joint moment data, when analyzed for all the fingers of the hand, provides insight into how strongly the muscle acts on each finger, thus providing insight into any possible deficiencies in the electrically induced grasp. This means that a better grasp template can be derived so that the neuroprosthesis user can get the most functionality out of their hand that is possible. This information also allows us to better utilize the intrinsic muscles to provide improved finger extension without overcompensating for the finger deficiencies, which could produce problems over time.

Plans for Next Quarter

During the next quarter, more focus will be placed on the evaluation of the intrinsic muscles in the hand grasp to provide improved finger extension. This will include measurements of joint angle and contact and grip force with and without the intrinsic muscles. In addition, effort will be placed into the evaluation of alternative methods of command-control to allow for an intuitive and easy control of the bimanual system.

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2. b. iv CONTROL OF HAND AND WRIST

Abstract

We simulated performance of a feedforward neural network controller for isolating the control of hand grasp and wrist posture. Wrist angle and hand grasp opening/contact force could be specified accurately and independently with a neural network feedforward controller that accounted for the interactions between hand and wrist muscles at the wrist. However grasp had a significant effect on wrist angle when using networks that did not account for the interactions. Gravity also had a very strong effect on the feedforward control indicating that its effects must be incorporated into the feedforward controller design, or that other compensation techniques such as feedback control will have to be used.

Purpose

The goal of this project is to design control systems to restore independent voluntary control of wrist position and grasp force in C5 and weak C6 tetraplegic individuals. The proposed method of wrist command control is a model of how control might be achieved at other joints in the upper extremity as well. A weak but voluntarily controlled muscle (a wrist extensor in this case) will provide a command signal to control a stimulated paralyzed synergist, thus effectively amplifying the joint torque generated by the voluntarily controlled muscle. We will design control systems to compensate for interactions between wrist and hand control. These are important control issues for restoring proximal function, where there are interactions between stimulated and voluntarily controlled muscles, and multiple joints must be controlled with multijoint muscles.

Report of progress

Interaction vs. Non-interaction Networks

In the last Quarterly Progress Report, we described the design of a feedforward controller to provide isolated control of hand grasp and wrist movement in the hand grasp neuroprosthesis. Briefly, the controller consists of two stages: a coordination network and an interaction network (Figure 2.b.iv.1). The function of the coordination network is to specify the grasp and wrist parameters based on subject input. The function of the interaction networks is to specify the activation levels of the hand and wrist muscles based on the biomechanical interactions between the hand and wrist. Computer simulations with a biomechanical model of the forearm and hand [Esteki and Mansour 1996; Lemay and Crago 1996] revealed that the muscle activation levels selected from the feedforward controller reduced the interaction between hand grasp and wrist movement.

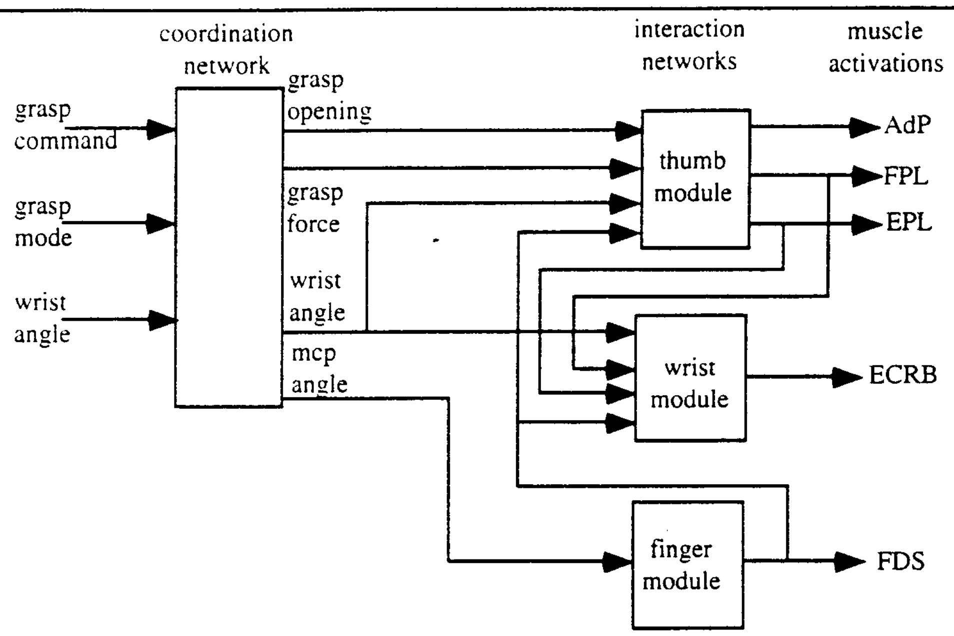

Figure 2.b.iv.1 Feedforward controller for isolating control of wrist and hand grasp. The coordination network specifies hand and wrist parameters in terms of grasp force, grasp opening, wrist angle, and finger angle. The network can be implemented as grasp templates similar to the ones used in this center. The interaction networks are divided into three modules. The inputs to the modules are the primary input the module is controlling, and secondary inputs that affect the primary inputs. The outputs of the modules are the muscle activation levels that control the primary input and minimize the biomechanical interactions between the primary and secondary inputs. Muscle abbreviations: extensor carpi radialis brevis (ECRB); flexor digitorum superficialis (FDS); extensor pollicis brevis (EPL); flexor pollicis longus (FPL); adductor pollicis (ADP).

To assess the reduction in the interaction between the hand and wrist, the performance of the feedforward controller was compared to a controller that does not compensate for biomechanical interactions (i.e. non-interaction networks). The non-interaction networks were the same design as the interaction networks, except the inputs not controlled by the networks (i.e. secondary inputs) were fixed to constant values. For example, the non-interaction wrist module had the finger and thumb inputs fixed to constants, while the desired wrist angle varied. Since performance will depend on the choice of constants, two different parameter sets were tested. In the first set, constant parameters were chosen for the posture of 4° wrist extension and 5.2 cm grasp opening. In the second set, the constant parameters were chosen for the posture of 20° wrist extension and 1.6 N grasp force. In both sets, the FDS activation was set at 0.10.

An example of simulated lateral grasp/wrist templates generated with the interaction and non-interaction networks is shown in Figure 2.b.iv.2. The templates specified a constant wrist angle of 15° extension as grasp opening and force changed with grasp command. Both set of constant parameters were used for the non-interaction networks. Notice that for the non-interaction networks, the ECRB activation level was constant regardless of changes in grasp opening or force. This resulted in large errors between the desired and actual wrist angle. For the interaction networks, the ECRB activation level varied as grasp opening and force changed, resulting in a wrist angle very close to the desired value of 15° extension. In a similar manner, the EPL activation level generated by the non-interaction networks was constant between 0% and 10% grasp command. However, since FDS activation was less than 0.10 in this range, grasp opening was greater than the desired value (since FDS activation decreases grasp opening). With the interaction networks, the EPL activation changed along with the FDS activation in order to maintain a constant grasp opening. Neither grasp force or MCP angle were significantly affected by the removal of the interaction effects. These same type of results were seen with a tenodesis template (grasp closure with wrist extension).

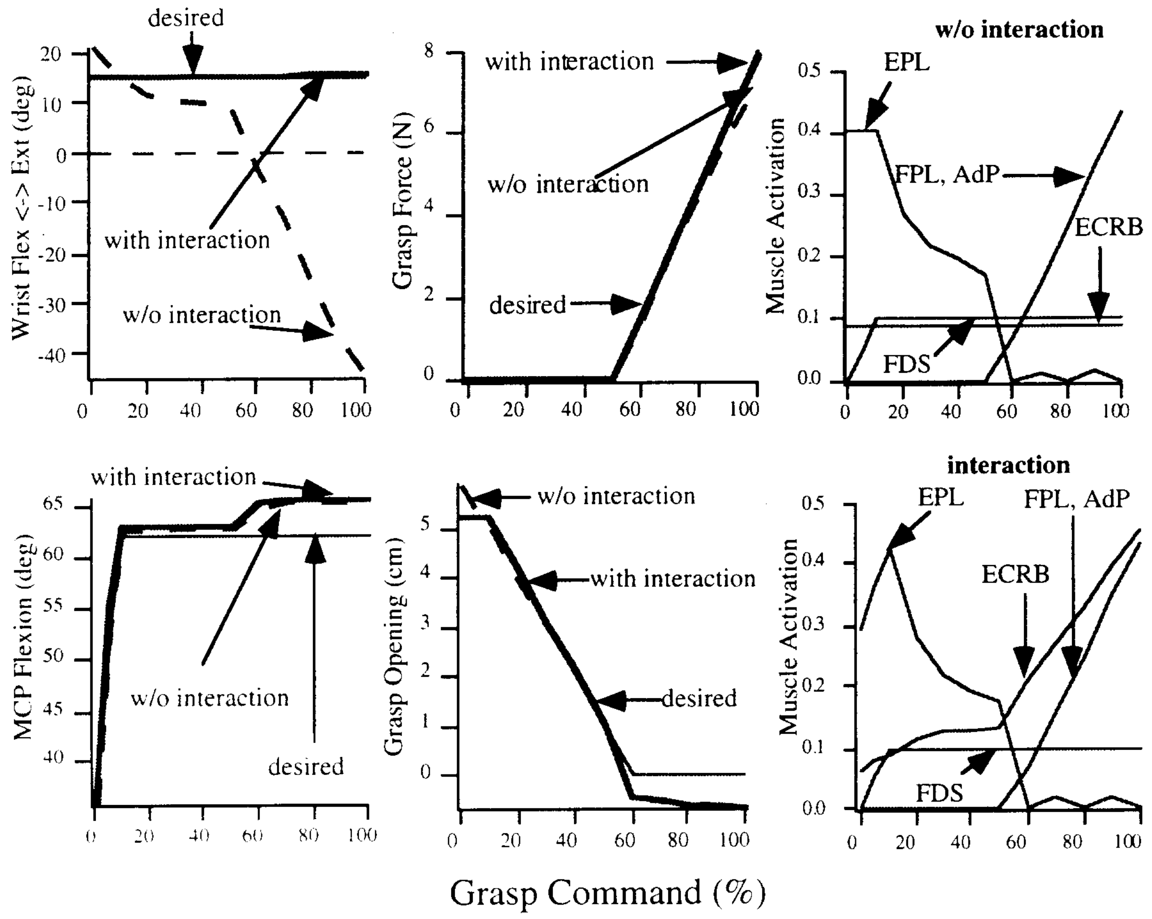

Figure 2.b.iv.2a Left four graphs: desired and simulated hand and wrist outputs vs. grasp command. Non-interaction networks set @ 5.2 grasp opening when wrist is at 4° extension. Light lines: desired value; Dark solid lines: output from interaction networks; Dark dashed lines: output from non-interaction networks. Right two graphs: the muscle activation maps generated with the non-interaction (top) and interaction (bottom) networks.

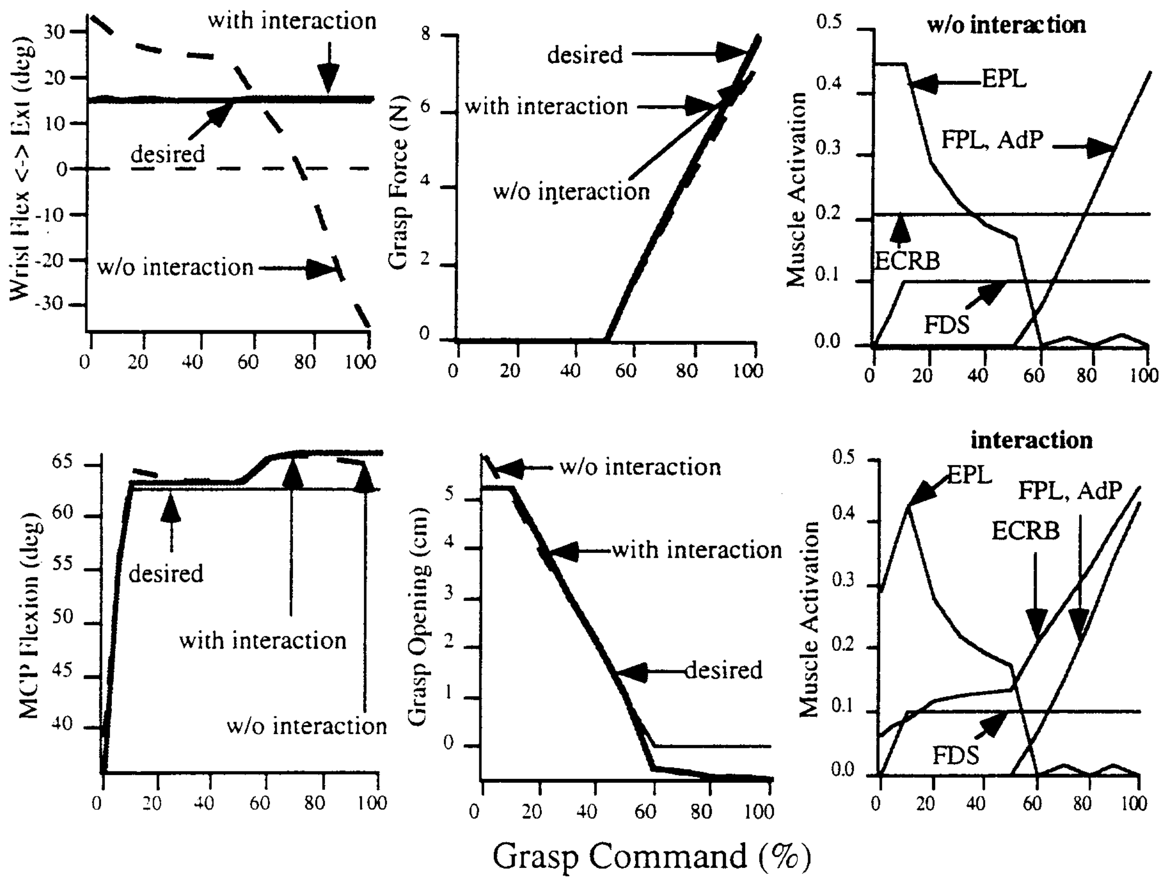

Figure 2.b.iv.2b Left four graphs: desired and simulated hand and wrist outputs vs. grasp command. Non-interaction networks set @ 1.6 N when wrist is at 20° extension. Light lines: desired value; Dark solid lines: output from interaction networks; Dark dashed lines: output from non-interaction networks. Right two graphs: the muscle activation maps generated with the non-interaction (top) and interaction (bottom) networks.

Effect of Gravity on Controller Performance
The training and testing of the feedforward controller up to this point consisted of no disturbances. However, the controller must perform adequately in the presence of disturbances in order to be of any practical use. Figure 2.b.iv.3 displays the performance of the feedforward controller in the presence of gravity (i.e. arm pronation and supination). The desired simulated template was a tenodesis lateral grasp. When the arm was neutral, the errors between the desired and actual grasp/wrist templates were insignificant. However, pronating or supinating the arm had a significant effect on wrist angle. Pronating the arm resulted in wrist flexion between 0% and 90% grasp command, while arm supination resulted in a large extension angle for the entire grasp command. Also, since pronating and supinating the arm reduces the ulnar moment at the wrist, grasp opening decreased. Gravity had a small effect on grasp force and MCP angle.

The effects of gravity seen here are much larger than reported in our last Quarterly Progress Report. The effects reported earlier were incorrect, since the wrong value for the gavitational constant was used. The differences are quite significant, and where we might have concluded from our previous results that gravity was not a significant effect, the present results would lead us to conclude that gravity is a serious factor that may limit the utility of the feedforward control scheme. However, there are some possible ways to counteract the effects of gravity. The first is to add an additional input to the feedforward controller. The input might come from an accelerometer on the forearm, and provide a signal proportional to the gravity effect. Thus, a feedforward controller might still be able to predict and compensate for the effects of gravity. A second possibility is to increase cocontraction to increase stiffness. However, this would limit the maximal wrist extension strength. The third possibility would be to add closed-loop feedback control. This will require a wrist angle sensor. Thus, the first solution is preferred.

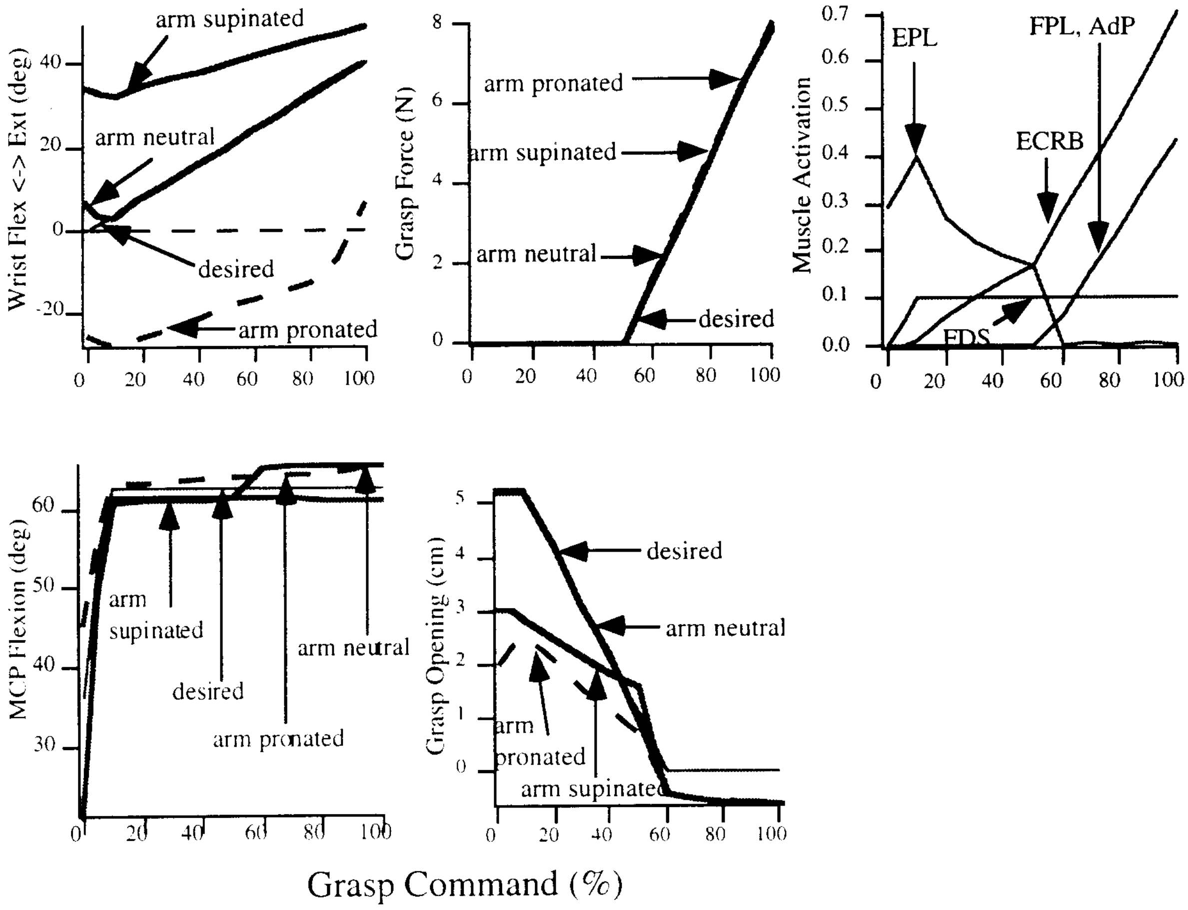

Figure 2.b.iv.3 Left four graphs: effect of gravity on wrist and hand grasp outputs. The arm was either neutral (position during training), pronated, or supinated. Right hand graph: muscle activation levels from the interaction networks. Except at 100 % command, the wrist could not extend beyond 0° when the arm was pronated.

Plans for next quarter

The interaction networks significantly reduced the interaction between hand grasp and wrist movement compared to the non-interaction networks. However, the effect of gravity on wrist angle is a potential problem in providing independent control of hand grasp and wrist movement with this control design. Ways to compensate for gravity that will be investigated include cocontraction of a wrist extensor and wrist flexor, and the addition of arm orientation and voluntary input to the feedforward control scheme.

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Lemay, M.A., and Crago, P.E., "A dynamic model for simulating movements of the elbow, forearm, and wrist", J. Biomechanics, 29: 1319-1330, 1996.